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Ionic Polymer-Metal Composites (IPMC) As Biomimetic Sensors and Actuators-Artificial Muscles

M. Shahinpoor[(a)], Y. Bar-Cohen[(b)], T. Xue[(b)], J.O. Simpson[(c)] and J. Smith[(c)]
(a)-Artificial Muscles Research Institute, University of New Mexico, Albuquerque, NM 87131, USA
(b)-NASA Jet Propulsion Laboratory (JPL), California Institute of Technology, Pasadena, Ca., 91109-8099
(c)-Composite and Polymer Branch, NASA Langley Research Center, Hampton, Va., 23681-0001

ABSTRACT

This paper discusses a number of recent findings in connection with ion-exchange polymer-noble metal composites (IPMC) as biomimetic sensors and actuators. These smart composites exhibit characteristics of both actuators and sensors. Strips of these composites can undergo large bending and flapping displacement if an electric field is imposed across their thickness. Thus, in this sense they are large motion actuators. Conversely by bending the composite strip, either quasi-statically or dynamically, a voltage is produced across the thickness of the strip between the two conducting electrodes attached. Thus, they are also large motion sensors. The output voltage can be calibrated for a standard size sensor and correlated to the applied loads or stresses. They can be manufactured and cut in any size and shape and in particular in the form of micro sensors and micro actuators for MEMS applications. In this paper first the sensing capability of these materials is reported by moving the tip end of a cantilevered sample and measuring the output voltage. The results were then plotted to get characteristic response of the composite for a given imposed tip displacement. The preliminary results shows the existence of a linear relationship between the output voltage and the imposed displacement for almost all cases. Furthermore, the ability of these ionic polymer-metal composites as large motion actuators and robotic manipulators is presented. Several muscle configurations are constructed to demonstrate the capabilities of these IPMC actuators. Further, the feasibility of using such smart composites as linear platform type actuators is discussed. A theoretical model is developed for such actuators and its predictions are compared to experimental results. This paper further identifies key parameters involving the vibrational and resonance characteristics of sensors and actuators made of IPMC's. When the applied signal frequency is varied, so does the displacement up to a point where large deformations are observed at a critical frequency called resonant frequency where maximum deformation is observed. Beyond which the actuator response is diminished. In this research paper, several samples of the actuators were made and tested with various dimensions to compare the vibrational behavior of the actuators. A data acquisition system was used to measure the parameters involved and record the results in real time basis. Finally reported in this paper are load characterization of such active polymer composites made with a noble metal such as platinum. A brief description of a proposed theory for this type of actuator was then discussed. The results showed that these actuators exhibit good force to weight characteristics in the presence of low applied voltages.

Keywords: Ionic Polymer-Metal Composite Sensor, Soft Actuator, Artificial Muscles, Biomimetic Sensor, Vibrations, Resonance.

2. INTRODUCTION

Ion-exchange polymer-metal composites (IPMC) are highly active actuators that show very large deformation in the presence of low applied voltage and exhibit low impedance. They operate best in a humid environment and can be made as a self-contained encapsulated actuators to operate in dry environments as well. They have been modeled as both capacitive and resistive element actuators that behave like biological muscles and provide an attractive means of actuation as artificial muscles for biomechanics and biomimetics applications. Grodzinsky[1], Grodzinsky and Melcher[2,3] and Yannas, Grodzinsky and Melcher[4] were the first to present a plausible continuum model for electrochemistry of deformation of charged polyelectrolyte membranes such as collagen or fibrous protein and were among the first to perform the same type of experiments on animal collagen fibers essentially made of charged natural ionic polymers and were able to describe the results through electro-osmosis phenomenon. Kuhn[5] and Katchalsky[6], Kuhn, Kunzle, and Katchalsky[7], Kuhn, Hargitay, and Katchalsky[8], Kuhn, and Hargitay[9], however, should be credited to have been the first investigators to report the ionic chemomechanical deformation of polyelectrolytes such as polyacrylic acid (PAA), polyvinyl chloride (PVA) systems. Kent, Hamlen and Shafer[10] were also the first to report the electrochemical transduction of PVA-PAA polyelectrolyte system. Recently revived interest in these area with concentration on artificial muscles can be traced to Shahinpoor and co-workers and researchers [11-14, 22-49], Osada[15], Oguro, Asaka and Takenaka[16], Asaka, Oguro, Nishimura, Mizuhata and Takenaka[17], Guo, Fukuda, Kosuge, Arai, Oguro and Negoro[18], De Rossi, Parrini, Chiarelli and Buzzigoli[19] and De Rossi, Domenici and Chairelli[20]. More recently De Rossi, Chiarelli, Osada, Hasebe, Oguro, Asaka, Tanaka, Brock, Shahinpoor, Mojarrad[11-69] have been experimenting with various chemically active as well as electrically active ionic polymers and their metal composites as artificial muscle actuators.

Essentially polyelectrolytes possess many ionizable groups on their molecular chain. These ionizable groups have the property of dissociating and attaining a net charge in a variety of solvent medium. According to Alexanderowicz and Katchalsky[17] these net charge groups which are attached to network of macromolecules are called polyions and give rise to intense electric fields of the order of 10^{10} V/m. Thus, the essence of electromechanical deformation of such polyelectrolyte systems is their susceptibility to interactions with externally applied fields as well as their own internal field structure. In particular if the interstitial space of polyelectrolyte network is filled with liquid containing ions, then the electrophoretic migration of such ions inside the structure due to an imposed electric field can also cause the macromolecular network to deform accordingly. Shahinpoor[18,22,25,26,28,29,31,32,33,34,35,36] and Shahinpoor and co-workers[21,23,24,27,30] have recently presented a number of plausible models for *micro-electro-mechanics* of ionic polymeric gels as electrically controllable artificial muscles in different dynamic environments. The reader is referred to these papers for the theoretical and experimental results on dynamics of ion-exchange membranes -platinum composite artificial muscles. Most Ion exchange polymeric membranes swell in solvents and by and large are hydrophilic. This gives rise to ability of the membrane to swell in water which can be controlled in an electric field due to ionic nature of the membrane. Furthermore by placing two electrodes in close proximity of the membrane walls and applying a voltage, the forced transport of ions within a solution through membrane becomes possible at microscopic level. For a solvent such as water then local swelling and deswelling of membrane can be controlled depending on polarity of the electrode nearby more like the behavior of the bimorphic materials. This can be achieved by chemical or other possible means of plating of conductive materials on membrane surfaces. Platinum is one such conductor that can be deposited inside the network and on the outer surfaces of IPMC. Also being ionic in microscopic structure, IPMC has the ability to shift its mobile ions of the same charge polarity within itself when it is placed in an electric field which results in ionic attraction or repulsion between the fixed charges of opposite polarity contained in the side groups within the polymer molecular chain. This leads to local collapse or expansion of the polymer membrane macroscopically. Physically this causes a stress gradient on opposite sides of the membrane causing it to bend. Therefore by applying an alternating signal at low voltage one can achieve membrane oscillation proportional to frequency and amplitude of the input signal. This bending oscillation can be utilized in various applications as in linear or platform type actuators.

The organization of this paper is such that it first discusses a number of recent findings in connection with ion-exchange polymer-noble metal composites (IPMC) as biomimetic sensors and actuators. These smart composites exhibit characteristics of both actuators and sensors. Strips of these composites will undergo large bending and flapping displacement if an electric field is imposed across their thickness. Thus, in this sense they are large motion actuators. Conversely by bending the composite strip, either quasi-statically or dynamically, a voltage is produced across the thickness of the strip between the two conducting electrodes attached. Thus, they are also large motion sensors. The output voltage can be calibrated for a standard size sensor and correlated to the applied loads or stresses. They can be manufactured and cut in any size and shape and in particular in the form of micro sensors and micro actuators for MEMS applications. These recent findings are described in this paper.

3-Biomimetic Sensing Capability of IPMC

Investigation of using ion-exchange-membrane metal composite (IPMC) materials as sensors can be traced to Sadeghipour, Salomon, and Neogi[21] where they used the composite membrane as a pressure sensor/damper in a small chamber which constituted a prototype accelerometer. Their research was involved with high frequency vibration sensing and suppression, and therefore much of their investigation involved higher bandwidth sensing.

In this paper the focus was on application of the IPMC sensor on quasi-static or dynamic displacement sensing where the response of the sensor against large imposed displacements was investigated. To get a better understanding of the mechanism of sensing, more explanation need to be given about the nature of the ionic polymers in general.

As shown in Figures 1 and 2, IPMC strips generally bend towards the anode and if the voltage signal is reversed they also reverse their direction of bending.

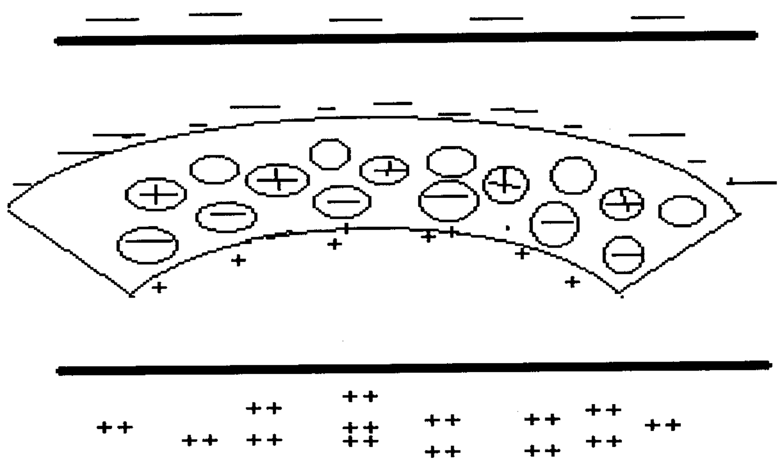


Figure 1. General redistribution of charges in ionic polymer due to imposed electric field.

Conversely by bending the material, shifting of mobile charges become possible due to imposed stresses. Consider Figure 2 where a rectangular strip of the composite sensor is placed between two electrodes. When the composite is bent a stress gradient is built on the outer fibers relative to neutral axis (NA). The mobile ions therefore will shift toward the favored region where opposite charges are available. The deficit in one charge and excess in other can be translated into a voltage gradient which is easily sensed by a low power amplifier.

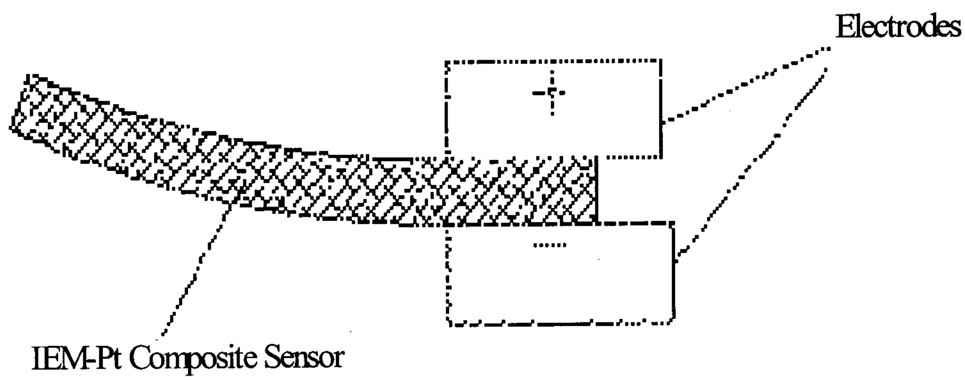


Figure 2. Simple IPPC sensor placed between two electrodes.

3.1-Quasi- Static Sensing

The experimental results showed that a linear relationship exists between the voltage output and imposed displacement of the tip of the IPMC sensor (Fig. 3). However the results were face sensitive while maintaining linearity, meaning the trend reversed itself when the sensor film was inverted (Fig. 4). This could be attributed to the fact that charges are built on the outer faces much like a capacitor which is then incrementally neutralized when the membrane sensor is inverted but same loading condition is applied.

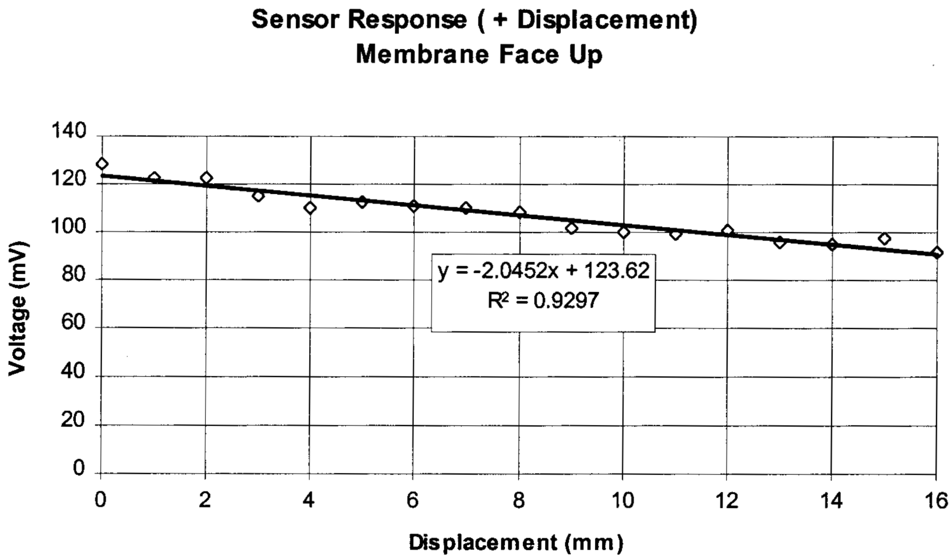


Figure 3. IPMC film sensor response for positive displacement input.

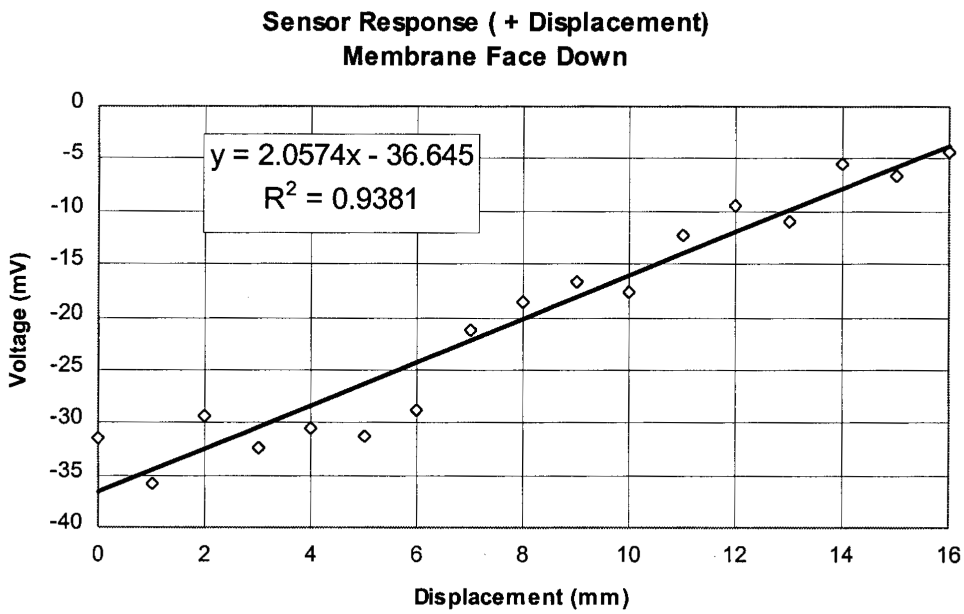


Figure 4. Inverted IPMC film sensor response for positive displacement input.

3.2-DYNAMIC SENSING

When strips of IPPC are dynamically disturbed by means of a dynamic impact or shock loading, nicely damped electrical response is observed as shown in Figure 5 below. The dynamic response were observed to be highly repeatable with a fairly high band width to 100's of HZ. This particular property of IPPC's may find a large number of applications in large motion sensing for a variety of industrial applications. Since these muscles can also be cut as small as one desires, they present a tremendous potential to micro-electro-mechanical systems (MEMS) sensing and actuation applications.

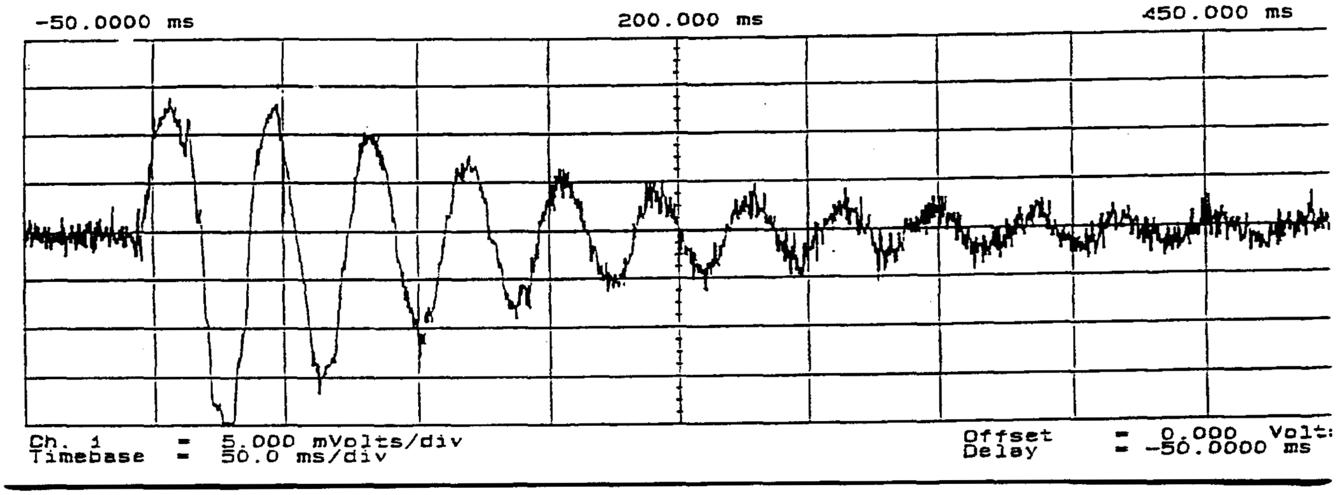


Figure 5- Dynamic sensing response in the form of output voltage of strips (40mmx5mmx0.2mm) of IPPC subject to a dynamic impact loading as a cantilever.

3.3- CONCLUSIONS

Remarkable linear behavior of the membrane was observed for each complete bending cycle. The low amplification factor of 17.3 with no signal conditioning proved to be sufficient for our sample in use. Finally the use of ionic polymeric metal composites such as IPMC as sensor may be useful where simplicity and low cost are sought. However more research needs to be done to find out the effect of chemical treatment on sensor output. Highly Dynamic sensing characteristics of IPPC strips were remarkable in accuracy and repeatability and were found to be superior to existing motion sensors and micro sensors.

4-Biomimetic Actuation Properties of IPMC's

4.1- General Considerations

Generally, actuators are used to operate robotic devices that include robotic arms, rovers, fingers and systems. Other applications include release mechanisms, antenna and instrument deployment, positioning devices, aperture opening and closing devices, real-time compensation for thermal expansion in space structures, biomedical devices, heart, muscles and circulation assist devices equipped with soft actuators. Increasingly, there are requirements to reduce the size, mass, and power consumption of actuation devices, as well as their cost. Electroceramics (piezoelectric and electrostrictive) offer effective, compact, actuation materials to replace electromagnetic motors. A wide variety of Electroactive ceramics (EAC) materials are incorporated into motors, translators and manipulators, in such devices as ultrasonic motors and inchworms. In contrast to electroceramics, IPMCs are emerging as new actuation materials with displacement capabilities that cannot be matched by the striction-limited and rigid ceramics [1]. Table 1 shows a comparison between the capability of IPMC materials and both electroceramics and shape memory alloys. As shown in Table 1, IPMC materials are lighter and their potential striction capability can be as high as two orders of magnitude more than EAC materials. Further, their response time is significantly higher that Shape Memory Alloys (SMA). The authors' current study is directed towards taking advantage of these polymers' resilience and the ability to engineer their properties to meet robotic articulation requirements. The mass producability of polymers and the fact that electroactive polymer materials do not require poling (in contrast to piezoelectric materials) help producing them at low cost. IPMC materials can be easily formed in any desired shape and can be used to build MEMS-type mechanisms (actuators and sensors). They can be designed to emulate the operation of biological muscles [2-5] and they have unique characteristics of low density as well as high toughness, large actuation strain constant and inherent vibration damping.

TABLE 1: Comparison of the properties of IPMC, SMA and EAC

Property	Ionic polymer-Pt Composites (IPMC)	Shape Memory Alloys (SMA)	Electroactive Ceramics (EAC)
Actuation displacement	>10%	<8% short fatigue life	0.1 - 0.3 %
Force (MPa)	10 - 30	about 700	30-40
Reaction speed	μsec to sec	sec to min	μsec to sec
Density	1- 2.5 g/cc	5 - 6 g/cc	6-8 g/cc
Drive voltage	4 - 7 V	NA	50 - 800 V
Power consumption	watts	watts	watts
Fracture toughness	resilient, elastic	elastic	fragile

4.2- DEVELOPMENT OF MUSCLE ACTUATORS

The IPMC muscle used in our investigation is composed of a Nafion® 117 (DuPont) film, i.e., a perfluorinated ion exchange membrane (IEM), which is chemically deposited platinum electrodes on its both sides. The thickness of the formed muscle actuator is 0.18-mm and it is cut in strips that is 1x0.125-inch in area and weighs 0.1 gram. To maintain the actuation capability the films need to be kept moist continuously. The commercially available Nafion has the following chemical formula,

$$
\begin{array}{l}
[-(CF_2\text{-}CF_2)_n\text{-}CF\text{-}CF_2\text{-}]_m \\
\qquad\qquad\quad | \\
\qquad\qquad\quad O\text{-}CF\text{-}CF_2\text{-}O\text{-}CF_2\text{-}SO_3\text{-}M \\
\qquad\qquad\qquad\quad | \\
\qquad\qquad\qquad\quad CF_3
\end{array}
$$

where $n \sim 6.5$, $100 < m < 1000$, and M^+ is the counter ion (H^+, Li^+ or Na^+). The structure and properties of the Nafion membranes have been the subject of numerous investigations (see for example [6]). One of the interesting properties of this material is its ability to absorb large amounts of polar solvents, i.e. water. Platinum, Pt, metal ions, which are dispersed through out the hydrophilic regions of the polymer, are subsequently reduced to the corresponding metal atoms. This results-in the formation of a dendritic type electrodes. In Figure 10, a scanning electron micrographs are shows in two magnifications, with an order of magnitude difference. On the left, a view is give of the edge of an electroded Nafion® muscle, the Pt metal covers each surface of the film with some of the metal penetrating the subsurface regions of the material. A closer view with x10 magnification is shown in Figure 6 on the right.

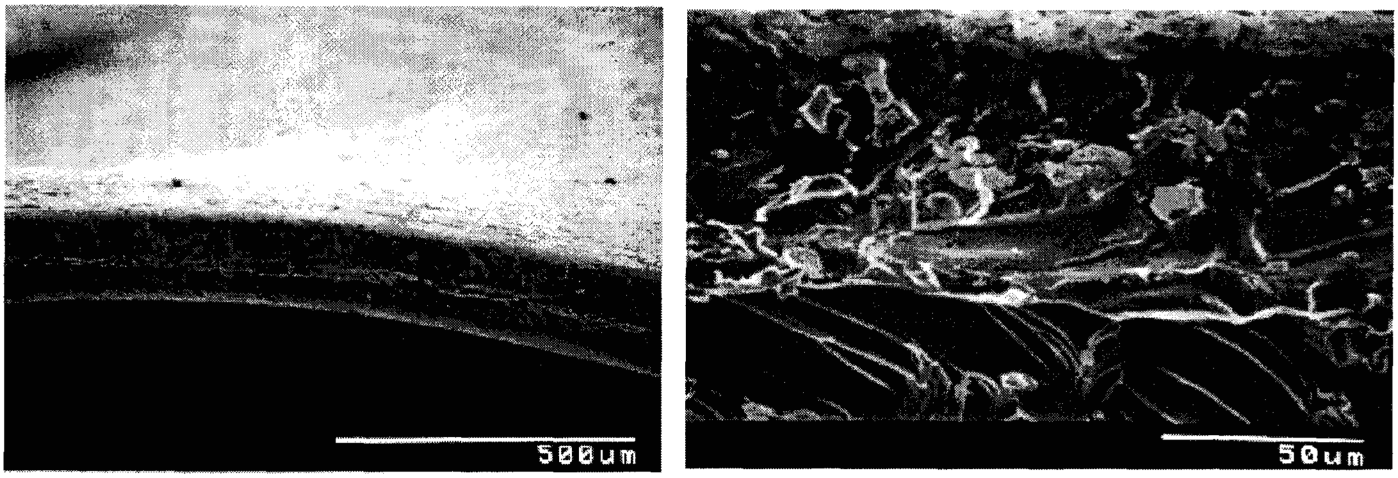


Figure 6: Scanning Electron Micrographs of the structure of Nafion®

When equilibrated with aqueous solutions these membranes are swollen and they contain a certain amount of water. Swelling equilibrium results from the balance between the elastic forces of the polymeric matrix and the water affinity to the fixed ion-exchanging sites and the moving counter ions. The water content depends not only on the hydrophilic properties of the ionic species inside the membrane but also on the electrolyte concentration of the external solution.

When an external voltage of 2 volts or higher is applied on a IPMC composite film, it bends towards the anode. An increase in voltage level (up to 6 or 7 volts) causes a larger bending displacement. When an alternate voltage is applied, the film undergoes movement like a swing and the displacement level depends not only on the voltage magnitude but also on the frequency. Lower frequencies (down to 0.1 or 0.01 Hz) lead to higher displacement (approaching 1 inch). Thus, the movement of the muscle is fully controllable by the applied electrical source. The muscle performance is also strongly dependent on the water content which serves as an ion transport medium and the dehydration rate gradient across the film leads to a pressure difference. The frequency dependence of the ionomer deflection as a function of the applied voltage is shown in Figure 7. A single film was used to emulate a miniature bending arm that lifted a mass weighing a fraction of a gram. A film-pair weighing 0.2-g was configured as a linear actuator and using 5V and 20 mW successfully induced more than 11% contraction displacement. Also, the film-pair displayed a significant expansion capability, where a stack of two film-pairs 0.2-cm thick expanded to about 2.5 cm wide (see Figure 8).

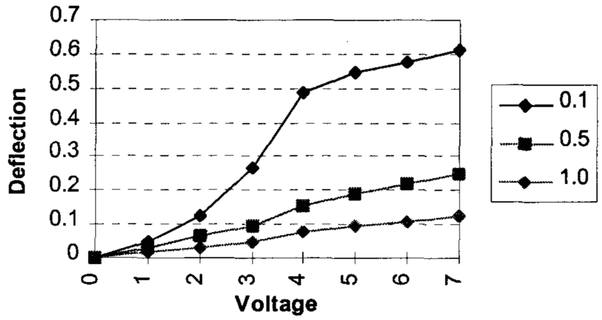


Figure 7: The deflection of a Nafion ionomer as a function of the frequency and the applied voltage.

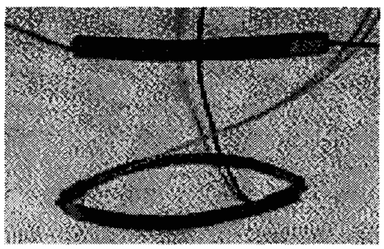


Figure 8: IPMC film-pair in expanded mode. A reference pair (top) and an activated pair (bottom).

4.3- MUSCLE ACTUATOR FOR ROBOTIC APPLICATIONS

IPMC films have shown remarkable displacement under a relatively low voltage drive, using a very low power. However these ionomers have demonstrated a relatively low force actuation capability. Since the IPMC composite films are made of a relatively strong material with a large displacement capability, we investigated their application to emulate fingers. In Figure 9, a gripper is shown that uses IPMC fingers in the form of an end-effector of a miniature low mass robotic arm.

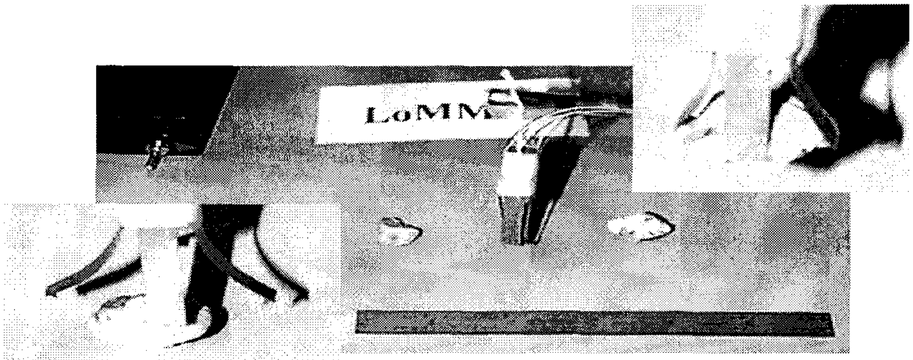


Figure 9: An end-effector gripper lifting 10.3-g rock under 5-V, 25-mW activation using four 0.1-g fingers made of perfluorinated ion-exchange membrane platinum composite.

The fingers are shown as vertical gray bars and the electrical wiring, where the films are connected back-to-back, can be seen in the middle portion of Figure 13. Upon electrical activation, this wiring configuration allows the fingers to bend either inward or outward similar to the operation of a hand and thus close or open the gripper fingers as desired. The hooks at the end of the fingers are representing the concept of nails and allow securing the gripped object that is encircled by the fingers.

So-far, multi-finger grippers that consist of 2- and 4-fingers were produced, where the 4-finger gripper shown in Figure 13 allowed to lift 10.3-g. This gripper prototype was mounted on a 5-mm diameter graphite/epoxy composite rod to emulate a light weight robotic arm. This gripper was driven by 5 volts square wave signal at a frequency of 0.1 Hz to allow sufficient time to perform a desirable demonstration of the capability of the Gripper -- opening the gripper fingers, bringing the gripper near the collected object, closing the fingers and lifting an object with the arm. The demonstration of this gripper capability to lift a rock was intended to pay the way for a future application of the gripper to planetary sample collection tasks (such as Mars) using ultra-dexterous and versatile end-effector.

To enhance the force actuation capability of IPMC composite actuators, techniques of producing thicker IPMC films are being developed. Further, we are seeking a better understanding of the actuation mechanism of ionomers as well as searching alternatives to Nafion as a base for ionomer actuators. Also, to protect the ionic constituents of IPMC composite films, encapsulation methods are being developed..

4.4- DESIGN OF LINEAR & PLATFORM TYPE ACTUATORS

For detailed dynamics description and analysis of the dynamic theory of ionic polymeric gels the reader is referred to Shahinpoor and co-workers (see the references). Since ionic polyelectrolytes are for the most part three dimensional network of macromolecules cross-linked nonuniformly, the concentration of ionic charge groups are also nonuniform within the polymer matrix. Therefore the mechanism of bending is partially related to migration of mobile ions within the network due to imposition of an electric field as shown in Figure 1. However, recent investigation by the author and hias co-workers point to a stronger effect due to surface charge interaction which will be reported later.

Figure 10 depicts the bending deformation of a typical strip with varying electric field, while Figure 11 displays the variation of deformation with varying frequency of alternating electric field.

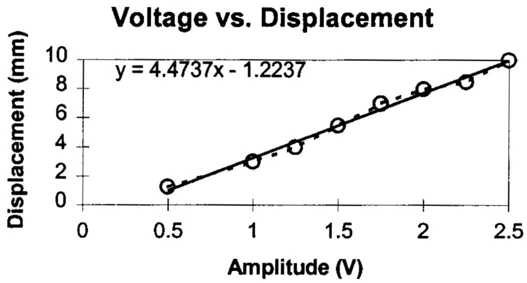


Figure 10-Bending Displacement versus Voltage for a typical strip of 5mmx0.20mmx20mm under a frequency of 0.5Hz.

Based on such dynamic deformation characteristics, linear and platform type actuators can be designed and made dynamically operational. These types of actuators are typically shown in Figures 11 and 12.

4.5-CONCLUSIONS

A new type of soft actuator and multi-fingered robotic hand were made from IPPC artificial muscles and were found to be quite superir to conventional grippers and multi-fingered robotic hands. The force capability, however, was observed to be still small while the robotic hands enjoyed large displacements with their multi-fingers. Encapsulation of the actuators to keep them moist appeared to be necessary for sustained actuation over a period of time. Efforts are underway to use the metal component in such composites to also play the role of encapsulants to keep the muscles moist and wet for long applications. Furthermore, In this section the feasibility of designing linear and platform type robotic actuators made with a polyelectrolyte ion exchange membrane-metal composite artificial muscle were presented. In order to achieve linear motion from these typically bending type actuators, a series of muscles made from ion-exchange-membrane-metal composites were cut in strips and attached either end-to-end or to one fixed platform and another movable platform in a cylindrical configuration. By especially prepared electrodes embedded within the platforms one can convert the bending response of each strip into linear movement of the mobile platform. By applying a low voltage the movement of free end of the actuator could be calibrated and its response could be measured, accordingly. A theoretical model was developed and was compared to experimental results.

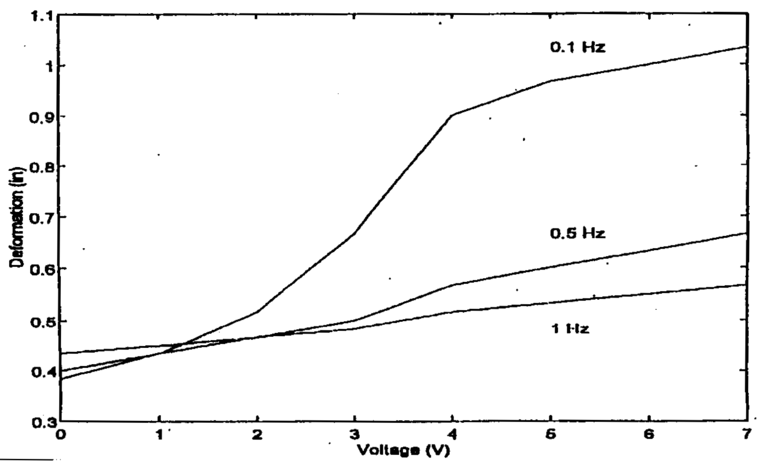


Figure 11-Frequency dependence of bending deformation of IPMC-Pt composite muscles

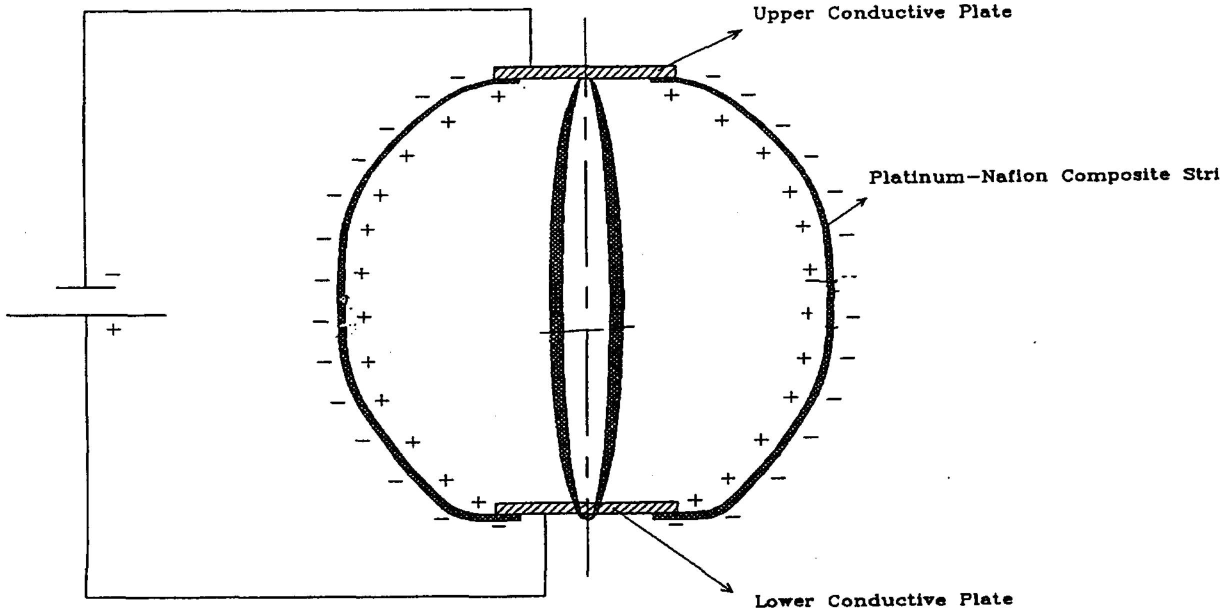


Figure 12- A typical linear-type robotic actuators made with IPMC-Pt composite legs

5-Large Amplitude Vibrational Response of IPMC's

5.1-General Considerations

Strips of polyelectrolyte Ion-Exchange membrane (IPMC) were used to study the large amplitude vibration characteristics of ion-exchange-membrane-metal composites. The ion-exchange membrane strips were chemically plated with platinum. A small function generator circuit was designed and built to produce a variety of approximately ±4.0V amplitude alternating wave at varying frequency. In order to study the feasibility of using ion-exchange-membrane-metal composite artificial muscles as vibration damper, a series of muscles made from ion-exchange-membrane-metal composites were cut in strips and attached either end-to-end or to one fixed platform and another movable platform in a cantilever configuration. By applying a low voltage the movement of free end of the beam could be calibrated and its response could be measured, accordingly. Typical data for the frequency-dependence of amplitude of lateral oscillations of the muscle strips subjected to alternating voltages of various forms such as sinusoidal, rectangular, saw-tooth or pulsed is presented. Furthermore, additional data is presented on static deformation of the strip with voltage as well as the frequency dependence of deflection-voltage curves.

5.2--EXPERIMENTAL OBSERVATIONS

A 15cmx15cm piece of IPMC (Nafion 117, Du Pont Company) was chemically deposited with platinum to produce the IPMC-platinum composite artificial muscle. Then typical strips of about 2-4cmx4-6mm of membrane composite was cut and completely swollen in a suitable solution such as water or alcohol to swell . The IPMC-Pt composite muscle strip typically weighed 0.1-0.4 grams and its thickness measured about 0.2mm after platinum was deposited on its two surfaces and was swollen in water. The strip was then held by a clamping setup between two platinum plate terminals which were wired to a signal amplifier and generator apparatus driven by Labview software through an IBM compatible PC containing an analog output data acquisition board. The amplifier (Crown model D-150A) was used to amplify the signal output of a National Instrument data acquisition card (AT-AO-10). A software was written to produce various waveforms such as sinusoid, square, triangular and saw tooth signals at desired frequencies up to 100 Hz and amplitudes up to 10 volts. When a low voltage was applied, the membrane composite bent toward the anode side each time. So by applying an alternating signal we were able to observe alternating bending of the actuator that followed the input signal very closely up to 35 Hz frequency. At voltages higher than 2.0 volts, the electrolysis of water in the composite was observed which led to degradation of displacement output of the actuator. Another factor affecting membrane composite performance was the dehydration. Water act as single most important element for the composite bending by sequentially moving within the composite depending on the polarity of the electrodes. The side facing the anode dehydrated faster than the side facing the cathode leading to a differential stresses which ultimately leads to bending of the composite. So, prior to each experiment, the composite was completely swollen in water. The displacement of the free end of a typical 2cmx4mm composite membrane was then measured for frequency range of 0.1-35 Hz for sinusoid input voltage at 2.0 volts amplitude (Figure 13). Resonance was observed at about 20 Hz frequency where the associated displacement was observed to be 7.5mm. It should be noted that as the actuator dehydrates the resonance frequency and maximum displacement varies accordingly. By encapsulating the strips in a plastic membrane such as Saran, the deterioration in the amplitude of oscillation decreased with time. However, the initial amplitude of oscillation for the same level of voltage was smaller than the unwrapped case due to increased rigidity of the strip. For our sample actuator the resonance occurred in the frequency range of 12 to 28 Hz for various swelling ratio.

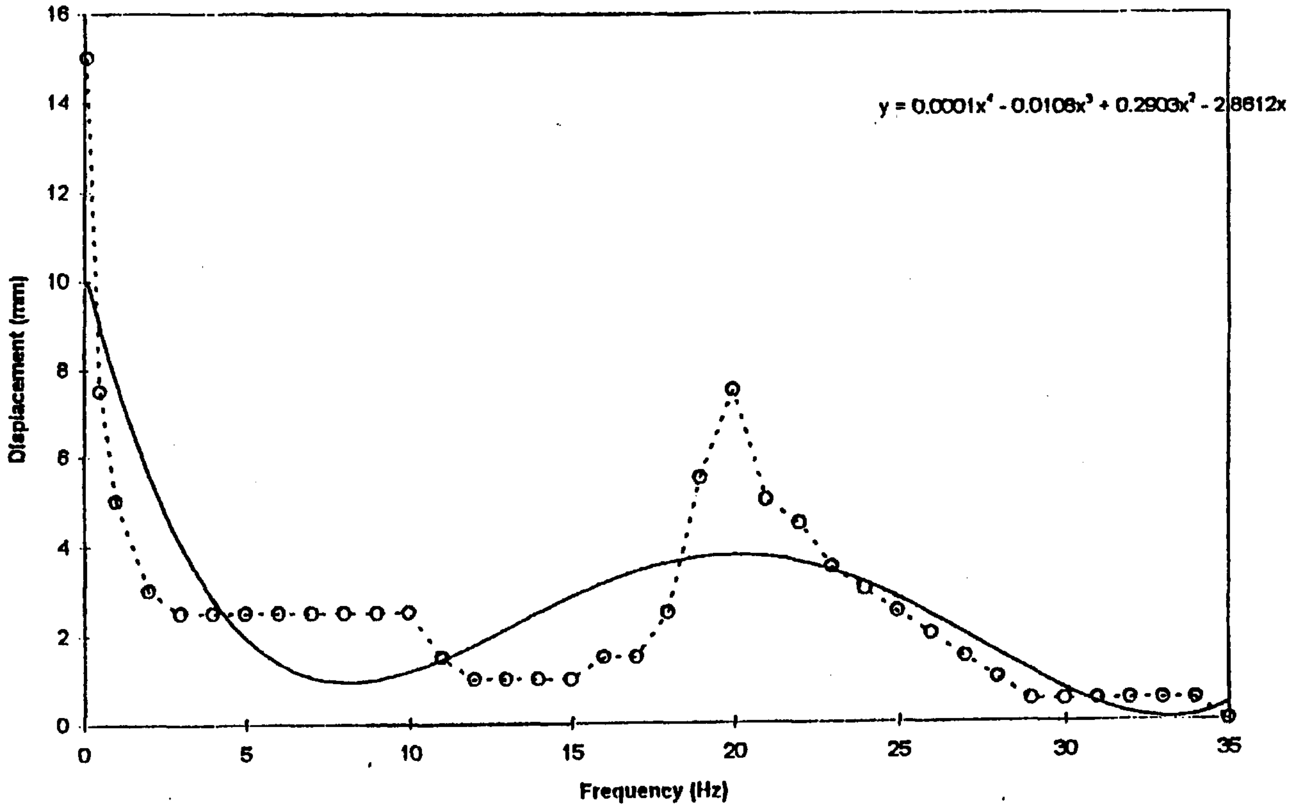


Figure 13- Amplitude of displacement versus the imposed frequency for a voltage of 2 volts for a 2cmx4mmx0.2mm sample

Based on such dynamic deformation characteristics, noiseless swimming robotic structures as shown in Figure 14 and cilia assembly-type robotic worlds, similar to coral reefs, as shown in Figure 15, were constructed and tested for collective vibrational dynamics. Furthermore, wing flapping flying machines, schematically shown in Figure 16, can be equipped with these muscles.

Figure 14. Robotic swimmer with muscle undulation frequency of 5 Hz (frame time interval, 1/3 second).

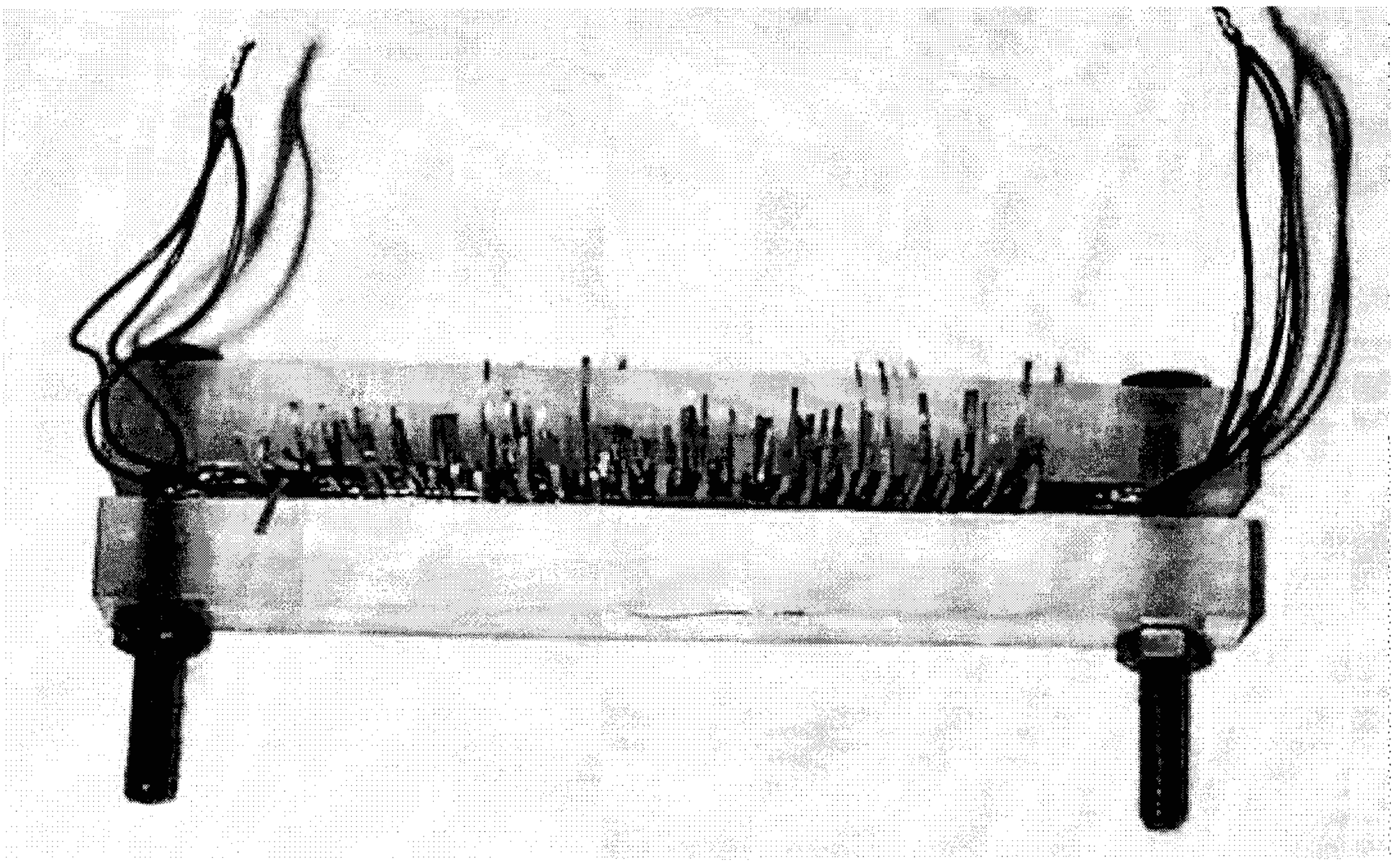


Figure 15--Cilia-Type assembly of IPMC-Pt Muscles Simulating Collective Dynamic Vibrational Response Similar to Coral Reefs and could create anti-biofouling surfaces

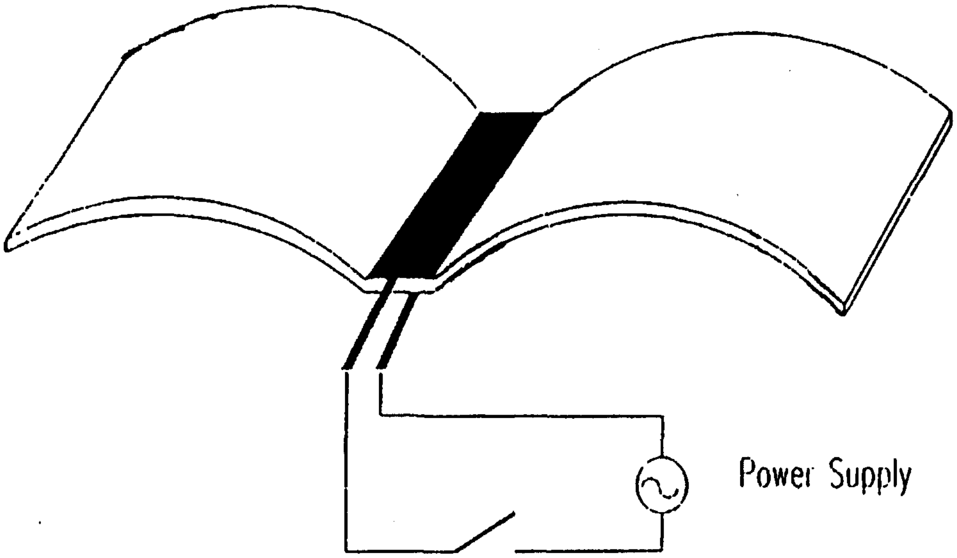


Figure 16-Wing-flapping flying machines design depicted schematically.

5.4- CONCLUSIONS

In this section the feasibility of designing dynamic vibrational systems of artificial muscles made with a polyelectrolyte ion exchange membrane-metal composite artificial muscle were presented. Our experiments confirmed that these types of composite muscles show remarkable bending displacement that follow input signal very closely. When the applied signal frequency is varied, so does the displacement up to a point where large deformations are observed at a critical frequency called resonant frequency where maximum deformation is observed. Beyond which the actuator response is diminished. Several samples of the actuators were made and tested with various dimensions to compare the vibrational behavior of the actuators. A data acquisition system was used to measure the parameters involved and record the results in real time basis. When a low voltage was applied, the membrane composite bent toward the anode side each time. So by applying an alternating signal we were able to observe alternating bending of the actuator that followed the input signal very closely up to 35 Hz frequency. At voltages higher than 2.0 volts, the electrolysis of water in the composite was observed which led to degradation of displacement output of the actuator. Another factor affecting membrane composite performance was the dehydration. Water act as single most important element for the composite bending by sequentially moving within the composite depending on the polarity of the electrodes. The side facing the anode dehydrated faster than the side facing the cathode leading to differential stresses which ultimately leads to bending of the composite. So, prior to each experiment, the composite was completely swollen in water. The displacement of the free end of a typical 2cmx4mm composite membrane was then measured for frequency range of 0.1-35 Hz for sinusoid input voltage at 2.0 volts amplitude. Resonance was observed at about 20 Hz frequency where the associated displacement was observed to be 7.5mm. It should be noted that as the actuator dehydrates the resonance frequency and maximum displacement varies accordingly. By encapsulating the strips in a plastic membrane (Saran) the deterioration in the amplitude of oscillation decreased with time. However, the initial amplitude of oscillation for the same level of voltage was smaller than the unwrapped case due to increased rigidity of the strip. For our sample actuator the resonance occurred in the frequency range of 12 to 28 Hz for various swelling ratio. Another measurement of the displacement of the free end was done by varying the amplitude of the sinusoid input voltage from 0.5-2.5 volts range at constant frequency of 0.5 Hz .

The observed remarkable vibrational characteristics of IPMC-Pt composite artificial muscles clearly point to the potential of these muscles for biomimetics applications such as swimming robotic structures, wing-flapping flying machines, slithering snakes, heart and circulation assist devices, peristaltic pumps and dynamic robotic cilia-worlds.

6-Load and Force Characterization of IPMC's

6.1-General Considerations

Polymeric ion-exchange membrane was acquired from commercially available source. This membrane is about 0.17mm thick and can be purchased as wide as one meter and any desired length. The membrane was then chemically cleaned and treated with platinum to form a composite that is active under electric field of low voltage. The thickness after chemical plating was about 0.2mm. The membrane was then fully hydrated in pure water bath. It was then cut in standard size of 20mm×5mm which was measured to be about 0.04grams in weight. A load cell (Transducer Techniques, model GS-30, 30 grams capacity) and corresponding signal conditioning module (Transducer Techniques, model TMO-1) together with a power supply was setup and connected to a PC-platform data acquisition and signal generation system composed of a 12-bit analog output board (National Instrument AT-AO-10) and a 16-bit multi-input-output board (National Instrument AT-MIO-16XE-50). A Nicolet scope was used to monitor the input and output waveform. Labview™ software was used to write a program to generate various waveform such as sinusoid, square, saw tooth, and triangular signals at desired frequency and amplitude (Fig. 3). Also a program was written to convert the output data from the load cell to force values and display on the monitor. The membrane actuator was then attached at one end to the load cell (load application point) and freed at the other end to be placed between two platinum electrode plates of 0.1mm thick (Aldrich Chemicals) which formed the jaws of a plastic forceps. Therefore the effective length of the membrane was 10mm when 10mm of the total length was placed between the electrodes. This made the effective weight of the muscle producing a force to be about 20 milligrams. A baseline was first established for each waveform with membrane actuator attached to load cell and electrodes to measure the initial pre-load and noise before actuation. Then a series of force data was generated by using a load cell correction factor of 0.255 and a constant frequency (0.5 Hz) signal input of 1.5, 2.0, 2.5, and 3.0 V rms amplitude voltages respectively. The resulting graphs were then adjusted for initial noise and pre-load and plotted over 5 seconds period (2.5 cycles). The force capability of these muscles , on the average was measured to be about 400 N/Kgm indicating that these muscles can lift almost 40 times their own weight. Figures 17 and 18 depict such general trends.

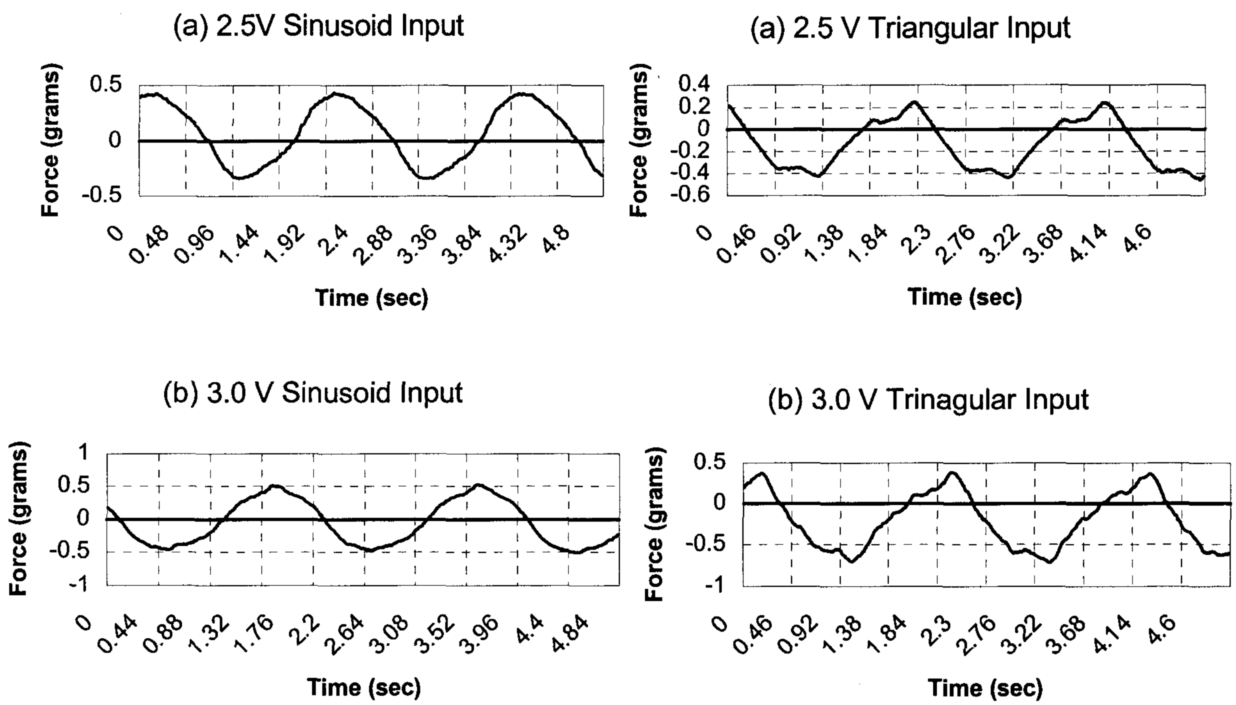


Figure 17. IPMC-Pt actuator response for sinusoid and triangular wave input at (a) 1.5, (b) 2.0, (c) 2.5, and (d) 3.0 Volts rms.

6.2- RESULTS AND DISCUSSION

The results showed that sinusoid and triangular wave form input to the IPMC-pt actuator generate more symmetric output forces, meaning they are relatively equal in either direction of travel (Fig. 17). The maximum forces were generated at higher input voltage amplitudes (3.0 V rms maximum) for most wave forms. The square and saw tooth input wave forms produced more nonuniform results in a sense that all or most of the generated forces were in one direction (Fig. 18). This can be attributed to the fact that there is insufficient time for ion travel to take place when the signal switches its polarity. However for the saw tooth signal, the voltage starts at zero and reaches a maximum in positive direction which results in force in one direction alone where in this case resulted in tension or downward movement of the IPMC-pt actuator. The next paragraph explains in details the effect of individual input waveform on the membrane actuator.

Sinusoid Input: This wave produced a uniform force in either direction of travel (tension and compression of the load cell). The output followed input best at 2.5 V amplitude. However the maximum forces were generated at 3.0 V input and reached 0.5 grams (10 times actuators mass) in either directions. This input shape appears to be more suitable for robotic controls where application of forces are involved for proper calibration and desired force response and command inputs.

Square Input: This wave generated the maximum output forces of all input wave forms reaching 1.25 grams (over 31 times actuators mass) in upward movement (load cell in compression) at 2.5 V amplitude. The reason for lower output at 3.0 V is not clear but square voltage causes sudden change of polarity which can attribute to rapid dehydration and heating of the membrane actuator all of which lead into increase in output in one direction only and may lead into actuator failure. At higher voltages, the generated forces appeared to shift toward negative (load cell in compression or upward movement).

Saw Tooth Input: This signal generated uniform forces in downward motion (load cell in tension) only reached a maximum of 0.65 grams (16 times actuators mass) at 3.0 V input.

Triangular Input: This wave form also produced a uniform force in both direction of travel and reached a maximum of 0.7 grams (17.5 times actuators mass) at 3.0 V in upward direction. However the best symmetry was observed to be at 1.5 V.

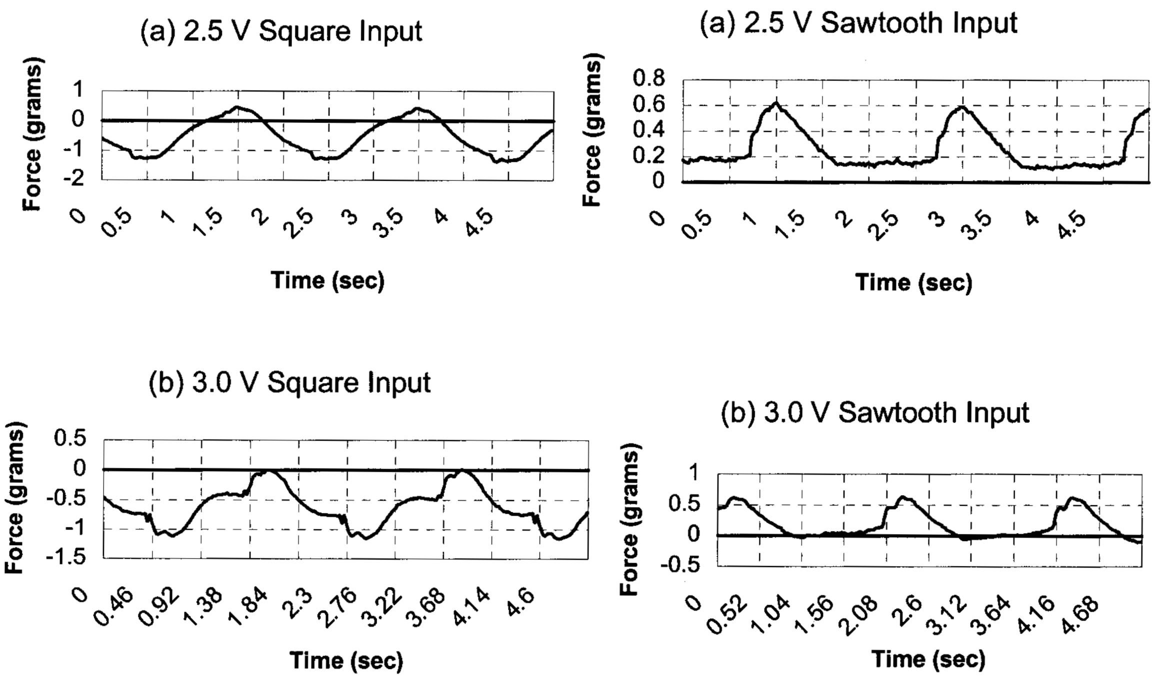


Figure 18. IPMC-Pt actuator response for square and saw tooth wave input at (a) 1.5, (b) 2.0, (c) 2.5, and (d) 3.0 Volts rms.

6.4- CONCLUSIONS

Polymer composite actuators such as IPMC-Pt proved to be practical for applications requiring large load and large motions as well as small load and micro-motion. These materials exhibit remarkable force to mass ratio (as high as 40 for these samples) and are inexpensive to manufacture. Depending on quality of manufacturing process control, actuators producing force in the order of over 50 times their mass have been observed previously in our laboratory. The experimental results showed that they are also sensitive to the shape as well as the amplitude of the applied input signal and result in different force responses accordingly.

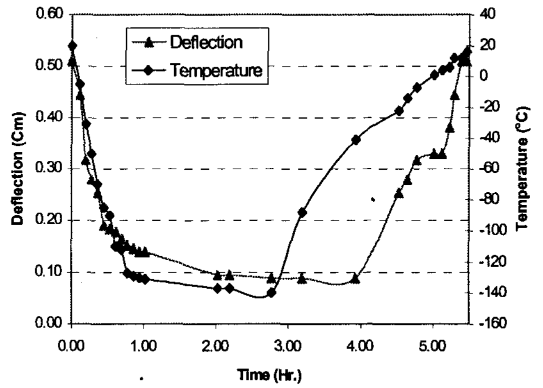
For robotic controls, sinusoidal input at low amplitudes produce more uniform response but at low magnitudes. This will enable a simple controls circuitry requirement to integrate the IPMC-Pt actuator in a robotic system such as gripper.

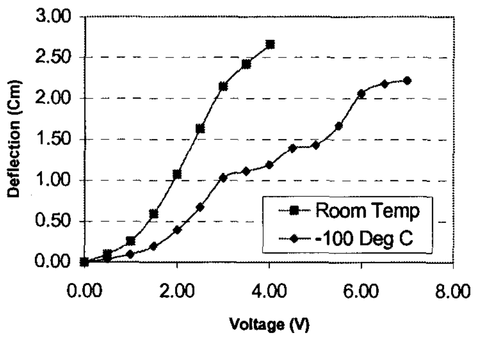
7-CRYOGENIC PROPERTIES OF IPMC ARTIFICIAL MUSCLES

In this section are reported a number of recent experimental results pertaining to the behavior of ionic polymer metal composites under low pressure (few Torrs) and low temperatures (-140 degrees Celsius). These experimental results have been obtained in a cryogenic chamber at JPL as well as a cryogenic chamber at the Artificial Muscles Research Institute at UNM. The interest at JPL was to study the actuation properties of these muscles in a harsh space environment such as one Torr of pressure and -140 degrees Celsius temperature. While at UNM the electrical properties , sensing capabilities as well as actuation properties of these muscles were tested in an atmospheric pressure chamber with a low temperasture of -80 degrees Celsius. The general results are that these materials are still capable of sensing and actuation in such harsh conditions as the following Figures 19 through 24 display. Furthermore, these IPMC artificial muscles become less conductive, i.e., their electrical resistance increases with decreasing temperature. This result appears to defy the generally accepted fact that resistance of metallic conductors increases /decreases with increasing/decreasing temperature, respectively.

Figure 19- Deflection characteristics of IPMC as a function of time and temperature

Figure 20- Power consumption of the bending actuator ionomer as a function of activation voltage.





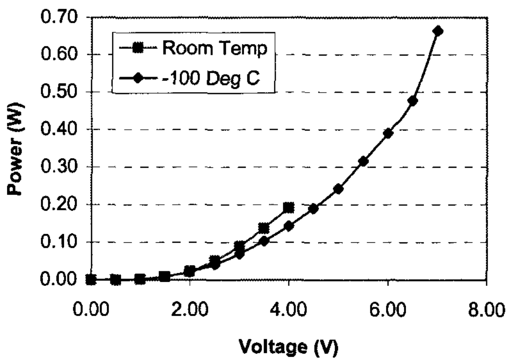
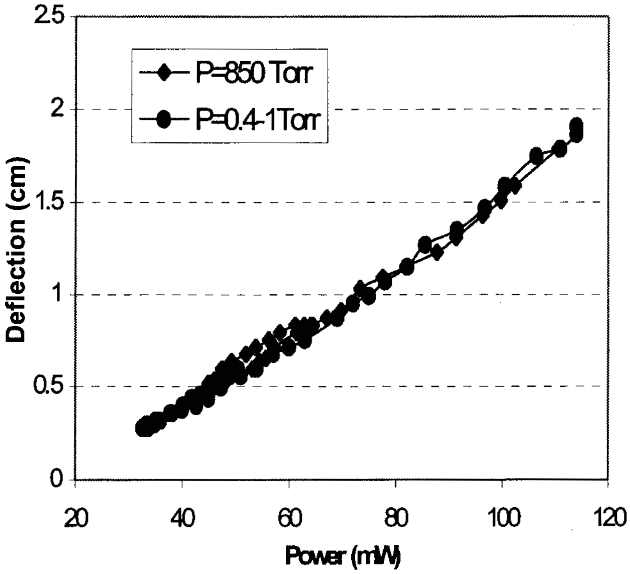
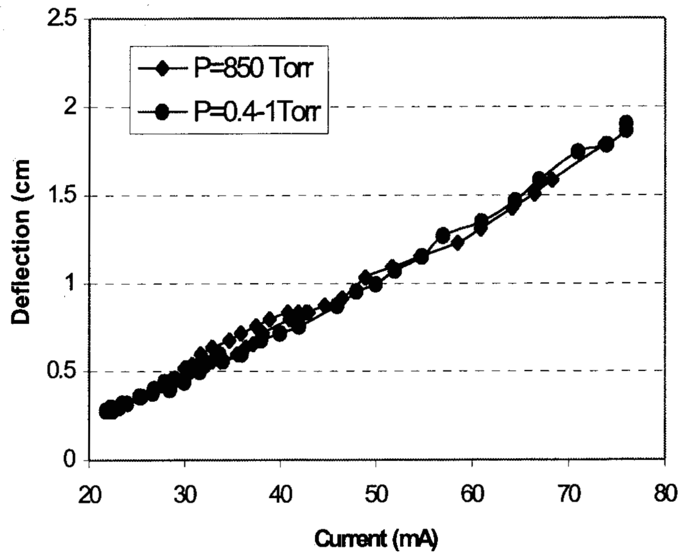


Figure 21- Deflection of the bending ionomer as function of voltage

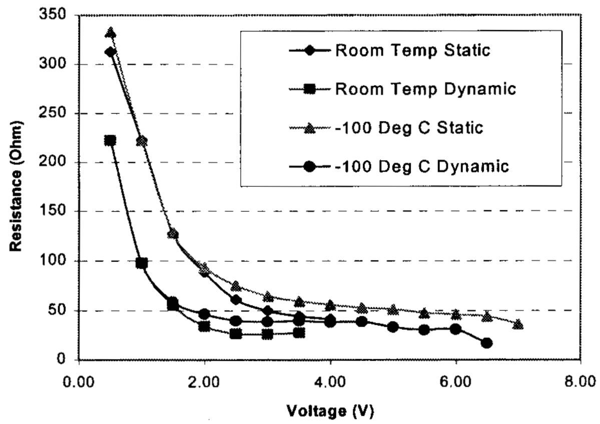


a. View of the deflection vs. power

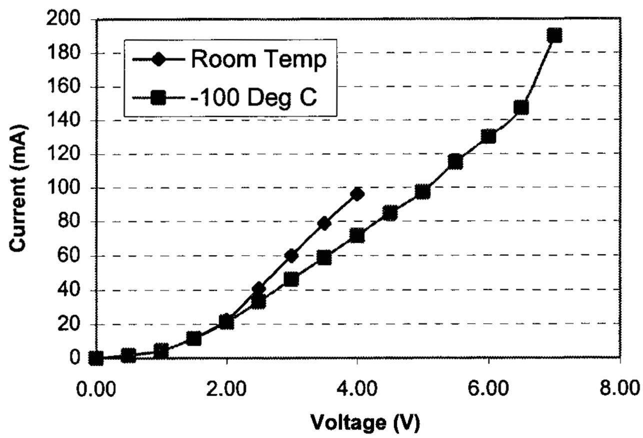


b. View of the deflection vs. current

Figure 22- Temperature changes in the range of 23 to -120°C versus power and current under a constant voltage of 3 volts And a frequency of 0.1-Hz.



Ionomer static (V/I) and dynamic (ΔV/ΔI) resistance at various temperature.



The relation between voltage and current for an ionomer that was exposed to RT and to -100°C.

Figure 23- Effect of temperature on the electrical resistance. Note: The graph is showing a greater resistance at lower temperatures. For any given temperature, there is a range of linear response of V vs. I, which indicates a close to a pure resistor response.

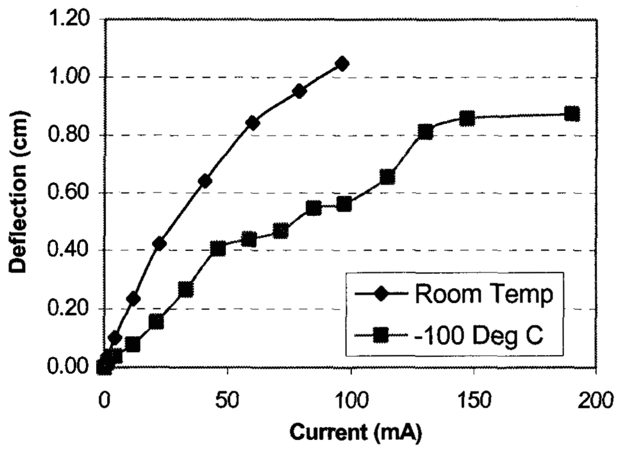


Figure 24-The relation between the current and the deflection for an ionomer that was exposed to room temperature and to -100°C.

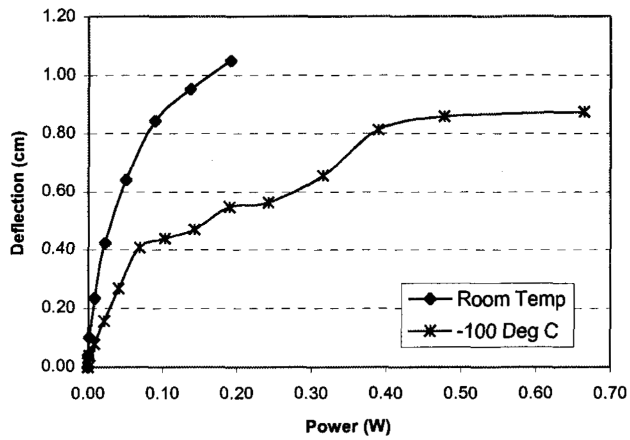


Figure 25The relation between the power and the deflection for an ionomer that was exposed to room temperature and to -100°C.

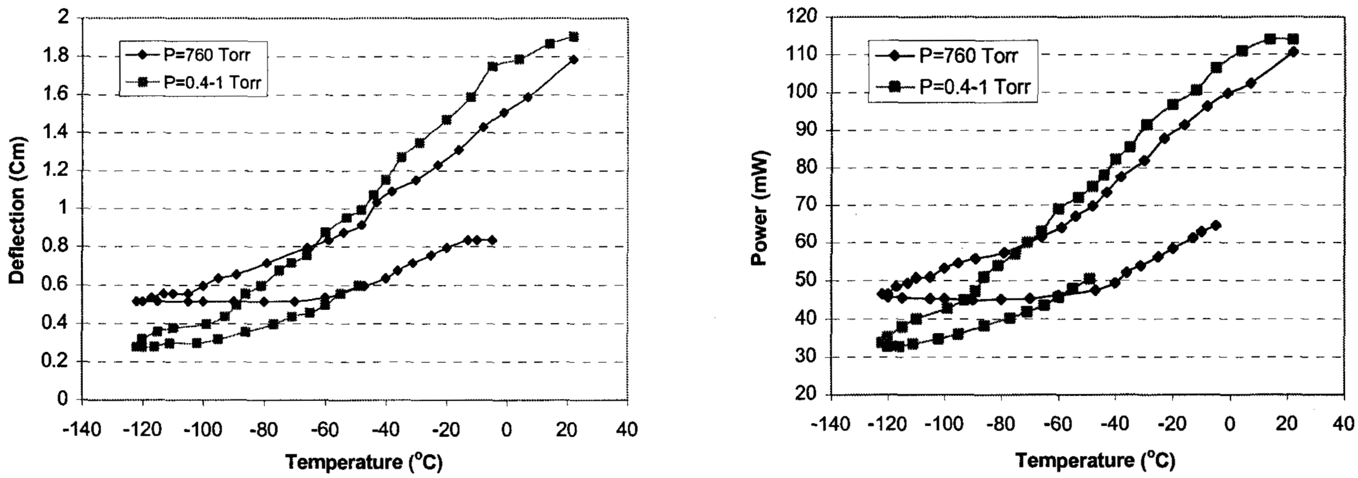


Figure 26-Deflection and power consumption of the ionomer muscle as a function of temperature with pressure as a parameter. Vpeak=3 V, Freq=0.1Hz.

8.0-ACKNOWLEDGMENT

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